INCREASING ELBOW TORQUE OUTPUT OF STROK PATIENTS BY EMG-CNOTROLLED EXTERNAL TORQUE

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Abstract- **A control algorithm for using homogenic EMG to control external assisting torque is developed for improving the elbow capability of stroke patients. The control signal to the manipulator is the difference between the weighted biceps and triceps EMG, so that the system moves with the forearm and provides assisting torque proportional to the voluntary effort. A nonlinear damping structure, mimicking physiological damping and incorporating the effects of cocontraction, was also included in the command for improving joint stability. The control algorithm has the advantage that the control is natural to the patient so that the learning process is simple. We tested the control algorithm in 5 normal subjects and 2 stroke patients. The results showed that the system could assist the subjects in completing work (tracking under loads) with less effort and without sacrificing performance.**
***Keywords* - Elbow, EMG, assisting torque, stroke**

I. INTRODUCTION

Hemiparesis, which means partial loss of muscle strength, is a common deficit in stroke patients and makes the patients less efficient in bearing loads in the affected side. Rehabilitation is the first choice for improving the muscle strength for patients with milder deficits, while functional electrical stimulation offers promise for patients with very severe deficits. For those patients with moderate deficits after rehabilitation, currently there was no adequate solution. We developed a system to increase the total torque capability of the elbow for this class of patients. The system was controlled by surface EMG of biceps and triceps, so that the manipulator arm moves together with the patient's forearm and provides assisting torque in proportion to the difference of weighted EMG of biceps and triceps. Contrary to the many attempts in the past that used EMG signal for switch control, in the current study, we used EMG signal for proportional torque control. We processed the EMG signal as proposed by Hogan [1] first, adaptively lowpass filtered the signal, then took the difference between the signals from biceps and triceps. The purpose of adaptive lowpass filtering is to perform low pass filtering while retain the fast change at the beginning of a movement. Using the difference of weighted EMG eliminated the problem of cocontraction when controlling the direction of movement, but it also eliminated the advantage of adjusting joint stiffness by cocontraction.

In order to adjust the joint stiffness according to the degree of cocontraction, we added a nonlinear damping as a function of the summation of biceps and triceps EMG signal and one third power of velocity [2] to the control command. The damping effect is relatively larger for small velocity and for larger cocontraction.

In this study, we investigated the performance of this assisting device in both normal subjects and stroke patients.

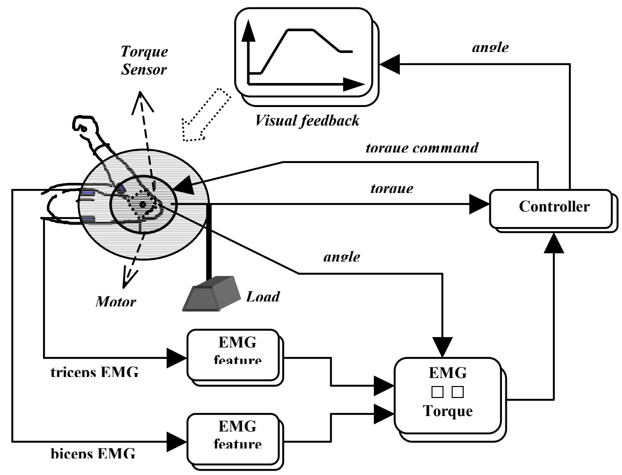

Figure 1. The schematic diagram of experimental setup

II. METHODOLOGY

Experimental setup

The overall experimental setup is shown in figure 1. The manipulator is set in position servo mode for measurement in isometric contraction and reaching experiments, and set in force servo mode for tracking experiments. The surface EMG of both biceps and triceps brachii are sampled in 500 Hz, bandpassed in 10 and 200 Hz and calculated as Hogan [1] proposed:

$$w_j = \left[\frac{1}{k} \left(\frac{1}{n} \sum_{1}^{n} m^2_{(j-1)n+i} \right)^{1/2} \right]^{1/b} \quad (1),$$

where j is an integer from 1 to n for numbering EMG signal, w_j is the jth transformed EMG signal, k and n are constants and m is the original sampled EMG signal. In the next step, the processed EMG is normalized:

$$M_j = \frac{w_j - w_{rel}}{w_{rel}}, \quad (2),$$

where w_{rel} is the processed EMG at the relaxed state. The estimated torque output of the system (T_{req}) is expressed as the difference between the weighted EMG of antagonists with an additional term representing nonlinear damping effect:

$$T_{req} = G \cdot (K_e \cdot M_e - K_f \cdot M_f - C_0 \cdot \left(M_e^2 + M_f^2\right)^{1/2} \cdot (\frac{dh_j}{dt})^{1/3}) \quad (3),$$

where G and C_0 are constants, K_e and K_f are gains of processed EMG of biceps and triceps brachii (M_e and M_f) and h_j is the joint angle. G is used to adjust the gain of this assisting device, deciding the ratio of torque produced by the

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Abstract

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manipulator and muscles. C_0 is used to adjust the contribution of nonlinear damping. Then, T_{req} is lowpass filtered with an adaptive filter [3]:

$$z_j \frac{T_{req\ j} - T_{req\ j-1}}{T} + T_{req\ j} = T_{com\ j} \qquad (4),$$

where T is the sampling period, T_{com} is the output of this filter and z is the variable time constant adjusted according to the following formula:

$$z_j = a \left| \frac{(p_j - p_{j-1}) \big/ T}{p_j} \right|^{-\frac{2}{3}} \qquad (5),$$

where a is a constant, P_j is the output of $T_{req\ j}$ going through a second order Butterworth lowpass filter. The purpose of this adaptive filter is to remove the noise amplified by difference operation, while maintaining quick response at the onset of a movement. We set a = 1, which allows the cutoff frequency to change in the range of 0.25 and 2.5 Hz. T_{com} is then fed to the manipulator.

Experimental procedure

Five normal subjects and two stroke patients participated in this study. The subject was in supine position with the forearm fixed to the manipulator. Both the desired and actual trajectories were shown on a monitor for visual feedback. Before the tracking experiments, isometric contraction measurements were performed to estimate K_e and K_f. The maximal voluntary flexion and extension torques are first measured. Since K_e and K_f were complex functions of torque amplitude and joint angle, we estimated these two constants at four joint angles (0, 45, 90 and 135 degrees, referring full extension as 0 degree) and three torque levels (30, 45 and 60% of maximal voluntary torque) in both flexion and extension directions. Separate maps of K_e and K_f as functions of torque and joint angle were constructed by interpolation and extrapolation.

Two sets of experiments were performed. Reaching experiments were designed to investigate the effects of nonlinear damping and adaptive filtering and performed only in the normal subjects. In this type of experiments, the subjects had to perform isometric step-up, hold and step-down exertion to the predefined force levels as the trajectory shown on the screen. The gain of the assisting torque was set to be 100%, which means the maximal voluntary and manipulator torque outputs are equal. We used integrated EMG magnitude (IEMG) and mean path length (MPL) as the performance indicators. IEMG, calculated as the mean of rectified EMG), was used to represent voluntary exertion and MPL, calculated as the ratio of total path length divided by the time spent between the onset of contraction to the settling down to the steady state, was used to estimate the smoothness of movement. One set of parameters (C_0=0.2 and a=2) was chosen for tracking experiments, which was designed to study the system performance.

In tracking trials, the target trajectory consisted of segments of ramp-up, hold and ramp-down movements. The load that the subject had to bear throughout the trial was set to be 40% of the minimum of maximal voluntary torques. The gain of the assisting torque was again set to be 100%. The tracking trials were repeated for the load in both flexion and extension directions.

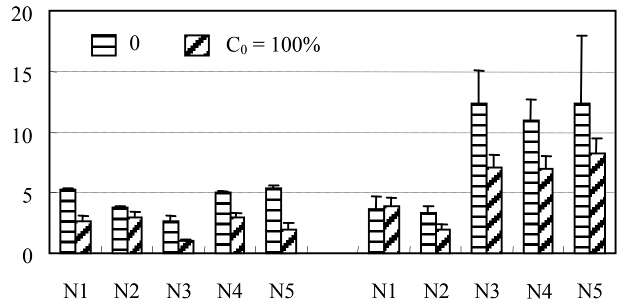


Figure 2. Agonist IEMG of normal subjects with and without assisting torque.

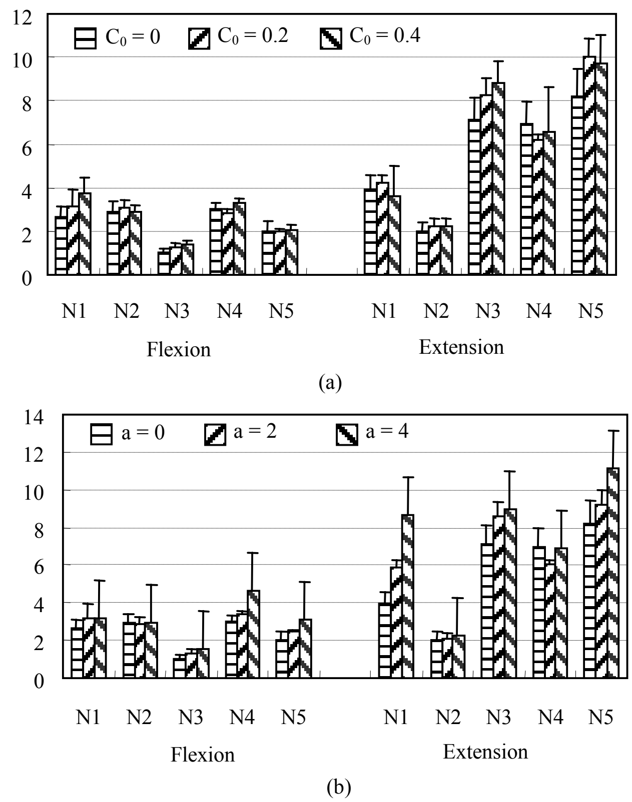
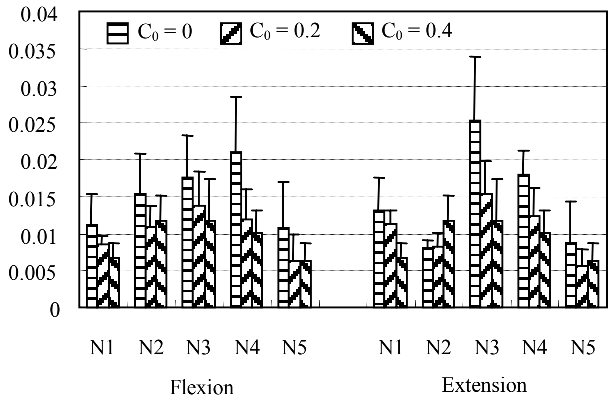


Figure 3. The effects of C_0 and a on agonist IEMG.

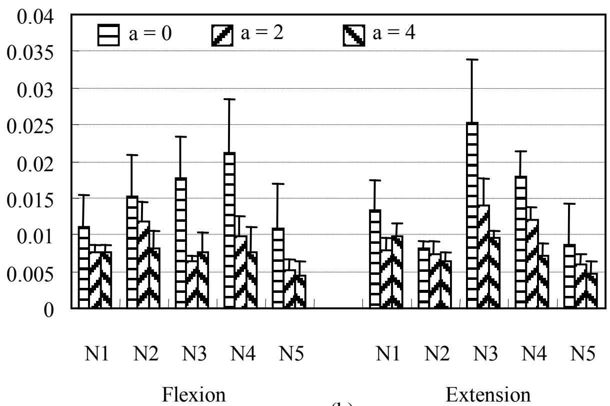
III. RESULTS

A. Results of reaching experiments

Fig. 2 shows the agonist IEMG with and without assisting torque, which clearly demonstrates that the assisting system decreases IEMG, i.e., the voluntary exertion with assisting torque. The subject can also feel the decrease in exertion subjectively. Fig. 3 shows the effects of C_0 and a on IEMG. C_0 has no clear effect on IEM on the chosen range, while IEMG increases with a. Fig. 4 shows the mean path length of the reaching experiments. Smaller value stands for smoother movements. As C_0 or a increases, the mean path length



(a)



(b)

Figure 4. Effects of C_0 and a on mean path length.

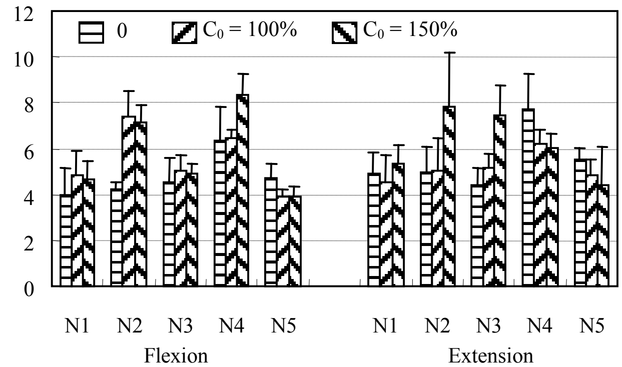
decreases. Yet, the improvement in smoothness pays price at the actual gain in torque capability, i.e., increasing the damping effect also increasing IEMG magnitude. Thus, as a compromise, we choose C0=0.2 and a=2 for the next tracking experiments.

B. Results of tracking experiments

We used root mean square error (RMS) and Integration of square of jerk (ISJ) as the performance indicators. Fig. 5 shows the results in normal subjects. The right group shows the results with the load in the flexion direction and the left group shows the results with the load in the extension direction. Except for subject N2 with loading in flexion direction, applying 100% of assisting torque does not impair the performance. As the assisting torque increases to 150%, 4/10 and 1/10 cases show performance impairment in RMS and ISJ, respectively. For stroke group, we only apply 100% of assisting torque. The results (Fig. 6) show that the performance is comparable in both with and without assisting torque.

IV. DISCUSSION

In the current study, we use the static EMG signal to construct the gain (K_e and K_f) maps. Though there were evidences that static and dynamic EMG may be different, the



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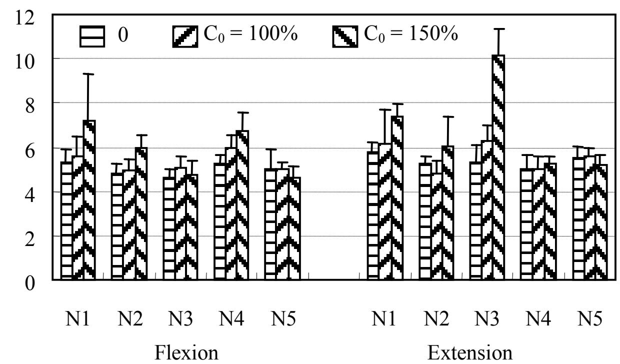
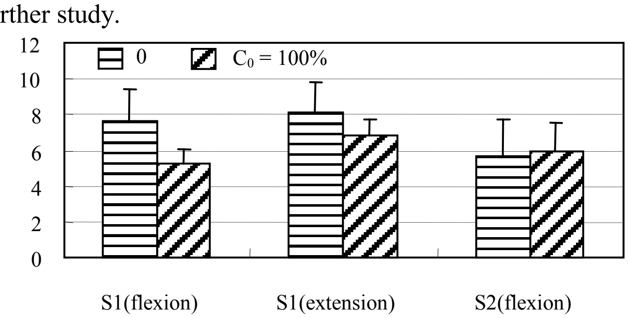


(b)

Figure 5. (a) RMS and (b) ISJ of normal subjects without and with 100% and 150% assisting torque

designed tracking movements are relatively slow. We think the designed velocity is sufficient for improving the elbow functions of stroke patients. The performance of the current control algorithm in movement with higher velocities needs further study.



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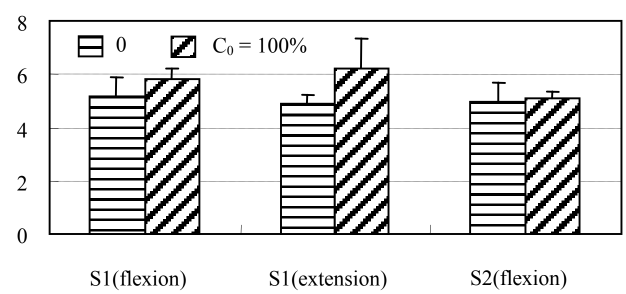

(b)

Figure 6. (a) RMS and (b) ISJ of affected side of stroke patients with and without assisting torque.

Since the magnitude of surface EMG is influenced by the shape of muscle, the contraction of the antagonist also changes the agonist EMG. We find that this effect is not negligible. Though we added a nonlinear damping structure which increases damping effect with cocontraction, we found in some subjects, cocontraction still leads to instability. We usually have to remind the subjects to relax and to reduce cocontraction to obtain better performance.

V. Conclusion

The presented results indicate that the developed control algorithm can increase the elbow torque output by 100% without reducing movement performance. Currently, we are investigating the system performance with assisting torque greater than 100% of voluntary muscular torque.

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